Journal of Electromyography and Kinesiology 20 (2010) 923–931

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Isometric torque–angle relationships of the elbow flexors and extensors in the transverse plane

Ilona J. Pinter *, Maarten F. Bobbert, A.J. "Knoek" van Soest, Jeroen B.J. Smeets

Research Institute MOVE, Faculty of Human Movement Sciences, VU University, Amsterdam, The Netherlands

ARTICLE INFO

Article history:
Received 2 October 2009
Received in revised form 30 March 2010
Accepted 3 May 2010

Keywords:
Maximal voluntary contractions
Optimal angle
Horizontal plane

ABSTRACT

Maximal voluntary isometric torque–angle relationships of elbow extensors and flexors in the transverse plane (humerus elevation angle of 90°) were measured at two different horizontal adduction angles of the humerus compared to thorax: 20° and 45°. For both elbow flexors and extensors, the torque–angle relationship was insensitive to this 25° horizontal adduction of the humerus. The peak in torque–angle relationship of elbow extensors was found at 55° (0° is full extension). This is closer to full elbow extension than reported by researchers who investigated this relationship in the sagittal plane. Using actual elbow angles during contraction, as we did in this study, instead of angles set by the dynamometer, as others have done, can partly explain this difference.

We also measured electromyographic activity of the biceps and triceps muscles with pairs of surface electrodes and found that electromyographic activity level of the agonistic muscles was correlated to measured net torque (elbow flexion torque: Pearson's $r = 0.21$ and extension torque: Pearson's $r = 0.53$). We conclude that the isometric torque–angle relationship of the elbow extensors found in this study provides a good representation of the force–length relationship and the moment arm–angle relationship of the elbow extensors, but angle dependency of neural input gives an overestimation of the steepness.

1. Introduction

In studies of motor control, arm movements are typically made in the transverse plane, i.e. the subject holds the arm at shoulder height, approximately at 90° elevation of the humerus, and endorotated such that the lower arm moves in the horizontal plane (Gomi and Kawato, 1997; Gribble et al., 2003; Kistemaker et al., 2006; Nijhof and Gabriel, 2006; Smeets et al., 1990). Working in this plane has the advantage that the effect of the gravitational force on the movement is eliminated. For simulation of this type of movement a model of the elbow joint and the muscles actuating it is needed, but an adequate description of isometric torque-angle curve of elbow extensors in the transverse plane is lacking in the present literature.

Most studies investigating torque-angle relationships at the elbow use a set-up in the sagittal plane, i.e. the subject holds the upper arm beside the thorax, approximately at 0° elevation of the humerus, and endorotated such that the lower arm moves in the anterior-posterior plane (Elkins et al., 1951; Osternig et al., 1977; Singh and Karpovich, 1966). We suspect that torque-angle curves obtained in the sagittal plane are inappropriate for modeling arm movements in the transverse plane because an important elbow flexor (m. biceps brachii) and an important elbow extensor (m. triceps brachii) are biarticular; their length is not only determined by the elbow angle but also by the angle of the glenohumeral joint. Also, the pronation and supination angle of the lower arm affects the maximal isometric torque development at the elbow (Elkins et al., 1951). A study that reports measurements of torque-angle relationships in the transverse plane and with the lower arm fixed in neutral position between pronation and supination is relevant because this corresponds to lower arm position in point-to-point movements used in studies of motor control.

A common method to obtain isometric torque-angle relationships is to ask participants to produce a maximal amount of force repeatedly at different joint angles. When using these maximal voluntary contractions (MVC) it is not guaranteed that participants activated their muscles maximally and/or to the same extent at all elbow angles. For elbow flexion torque it has been shown that healthy participants are unable to fully activate the biceps but that muscle inhibition is small (2%) (Dowling et al., 1994) and not related to elbow angle within a range of 30–120° (Brondino et al., 2002). For elbow extension torque this has not yet been investigated. Therefore, it seems relevant to measure electrical activity

* Corresponding author. Address: Research Institute Move, Faculty of Human Movement Sciences, VU University, Van der Boechorststraat 9, 1081 BT Amsterdam, The Netherlands. Tel.: +31 20 59 88452.
E-mail address: i.pinter@fbw.vu.nl (I.J. Pinter).

1050-6411/$ - see front matter © 2010 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jelekin.2010.05.001

(EMG) of biceps and triceps when obtaining isometric torque-angle relationships of the elbow. This can indicate whether variations in neural input of the agonistic muscle contributed to the shape of the torque-angle curves as found in the present study. Based on the above mentioned research we expect this contribution to be small (<5%). Furthermore, we wanted to estimate based on the antagonistic muscle activity how much we underestimated maximal isometric torque of the agonist by neglecting the contribution of the antagonistic muscle torque to the measured net joint torque.

Another relevant issue that we wanted to address in this study is the level of detail needed in modeling the shoulder girdle, an assembly of joints between thorax, scapula, clavicula and humerus, when it comes to simulating arm movements in the transverse plane. Describing how the maximum in the torque-angle curves of elbow flexors and extensors depends on horizontal adduction angle of the humerus will provide information on this point. We expect that with horizontal adduction the biceps (short head) would shorten and triceps (long head) would lengthen, causing the maximum in the torque-angle curve of these muscles to shift towards a more extended elbow angle. This could cause a similar shift in the total torque-angle curve of elbow flexors and extensors.

The purpose of this study is to gain more detailed knowledge of the isometric torque-angle relationships at the elbow. Maximal voluntary isometric torque of both elbow flexors and extensors was measured over a wide range of elbow angles and at two different prescribed horizontal adduction angles between thorax and humerus. The isometric torque-angle relationship of elbow extensors in the transverse plane was documented and its dependency on the horizontal adduction angle is investigated for both elbow extensors and flexors. We hypothesize that the maximum in the torque-angle curve of both elbow flexor and extensor will shift to more extended elbow angle with increasing horizontal adduction in the shoulder. With this study, we hope to contribute to the development and validation of musculoskeletal models of the elbow and shoulder to be used in simulating motor tasks that involve arm movements in the transverse plane.

2. Methods

2.1. Participants

Eleven healthy subjects (6 male and 5 female) with a mean age of 32 years (range 22–44 years) and without physical constraints at the neck, shoulder or arm at the time of the experiment volunteered. Participants varied in their professional activity and the sport they practiced in their leisure time. The local ethical committee approved the experiment. After receiving information about the experimental procedures, all participants signed an informed consent form.

2.2. Experimental set-up

Subjects were seated in the dynamometer and strapped tightly to the chair with safety belts to prevent trunk motion. The axis of rotation of the dynamometer arm was oriented vertically. The chair was adjusted so that the participant's arm was at shoulder height when placed on the dynamometer arm. For all participants the right arm was measured. The lower arm was fixed in neutral position in between pronation and supination. The length of the dynamometer arm was adjusted so that the rotation axis of the elbow (epicondylus medialis humeri) was in line with the rotation axis of the dynamometer arm.

Due to deformation of soft tissue, the joint angles changed from the values prescribed by the protocol when going from inactive state to maximal voluntary contraction. For this reason, we determined joint angles in the transverse plane at the moment of maximal voluntary contraction from a photograph taken during each trial. Passive markers were placed on the wrist (midway between the processus styloideus of the radius and that of the ulna), on the epicondylus lateralis humeri and on the most lateral edge of the acromion to facilitate detection of bony landmarks at the photographs.

Because the biarticular biceps and triceps originate at the scapula and not at the thorax, we also determined the angles in the transverse plane between thorax and scapula (φ_{ts}), scapula and humerus (φ_{sh}) in addition to the angles between thorax and humerus (φ_{th}) and humerus and ulna (φ_{elb}). See Fig. 1 for definitions of relative angles. We chose to measure the orientation of the scapula with an antenna containing two markers on either side, placed on the lateral part of the acromion and aligned with the spina scapulae. Previous research has shown that using a skin-based sensor at the acromion allows for 3D-tracking of the movement of the scapular bone, provided that the humerus elevation angle remains below 100° (Karduna et al., 2001; Meskers et al., 2007). We acknowledge that using a skin-based acromion marker for 2D-tracking leads to additional projection error due to scapular medial-lateral rotation and anterior-posterior tilt.

2.3. Equipment

Joint torque at the elbow was obtained with a Biodex System 3 dynamometer (Biodex Medical Systems, Inc., New York) at a sample frequency of 100 Hz. Electrical activity of biceps and triceps was collected using a Porti sytem (TMS International BV, Enschede, The Netherlands) operating at 2000 Hz. The static transverse-plane kinematics were captured with a photo camera (Nikon Coolpix 990: 3.34 megapixel) mounted overhead.

2.4. EMG

Two pairs of surface electrodes were placed, one pair on biceps brachii and one pair on the triceps brachii (lateral head) as de-

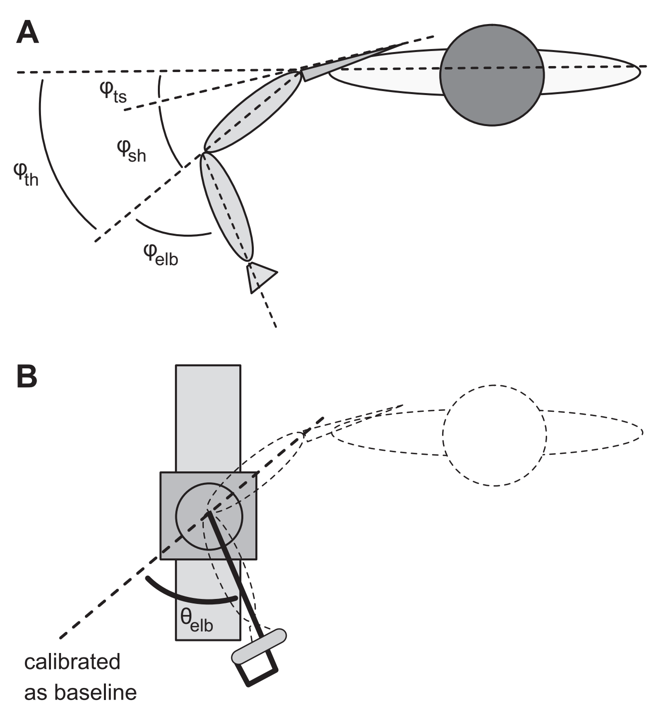

Fig. 1. (A) Definitions of relative angles between thorax and scapula (φ_{ts}), scapula and humerus (φ_{sh}), thorax and humerus (φ_{th}) and ulna and humerus (φ_{elb}) as used in this study. Note that with these definitions the angle is zero when the joint is maximally extended. (B) The angle imposed by the dynamometer is defined as θ_{elb}.

scribed on the website of the SENIAM project (http://www.seniam.org/), at a center-to-center inter-electrode distance of 2 cm. Electromyographic (EMG) data were filtered bidirectionally with a high-pass filter (Butterworth, cut-off 5 Hz) to remove possible movement artifacts, rectified, and then smoothed with a low-pass filter (effective time constant of 50 ms). The resulting smoothed and rectified EMG will be referred to as srEMG.

2.5. Experimental procedure

Maximal voluntarily isometric contractions were performed at 10 different elbow angles as set by the dynamometer (for imposed elbow angles (θ_{elb}) see Table 1) and at two different horizontal adduction angles between thorax and humerus (θ_{th}): 20° and 45°. A horizontal adduction angle of 45° was the maximal angle that could be achieved within the set-up of the dynamometer. Participants were instructed to build up either a flexion or an extension torque to maximum within a 3-s period. For each θ_{elb} one pair of contractions was performed: 3 seconds of flexion and 3 seconds of extension, separated by a 10-s relaxation period. Participants were then given a 20-s break after which a new θ_{elb} was set and a new pair of contractions was performed. After a short practice of three contraction pairs, two series of 10 contraction pairs (one contraction pair at each prescribed θ_{elb}) were performed. The two series differed in imposed horizontal adduction angle (θ_{th}) and will be referred to as TH20 and TH45. In between series, participants could relax as long as they needed. The two horizontal adduction angles, the 10 elbow angles, and starting with either flexion or extension in the contraction pair were presented in random order.

2.6. Data processing

From each contraction, a maximal torque value (T) was obtained by finding the highest average over a 0.5-s interval. To be able to compare data among subjects, we normalized these torques values for their average over all 20 contractions performed in the same direction.

The values for srEMG were calculated over the same 0.5 s intervals as used for averaging of the corresponding torque values. They were normalized in a similar way as the torque values: maximal biceps srEMG values were divided by their average over all twenty contractions of elbow flexion torque, and maximal triceps srEMG values by their average over all 20 contractions of elbow extension torque.

Angles between body segments (φ_{ts}, φ_{sh}, φ_{th} and φ_{elb}) during contractions were calculated using the photographs that were taken during the trials. The photographs were also used to check if the elbow rotation axis remained in line with the rotation axis of the dynamometer. We did not instruct participants to keep the elbow joint in place because we found it not desirable that participants would reduce their amount of force to follow this instruction. Instead, we estimated how much misalignment was allowed to keep relative error of the measured torque within 5% of the net torque around the elbow. For all contractions the misalignment remained within 5%-error margin. Details on the estimation of the relative error can be found in Appendix A.

2.7. Statistics

For each prescribed θ_{elb} we calculated mean and standard error of the mean (SEM) over all 11 subjects for the normalized torque values, the normalized srEMG values and the measured elbow angles (φ_{elb}). This leads to four mean T-φ_{elb} curves and their corresponding srEMG-φ_{elb} curves for the biceps and triceps. We report SEM instead of standard deviations because this study is concerned with providing mean T-φ_{elb} curves and the SEM is illustrative in how precise the reported mean curves are.

For every participant we modeled the T-φ_{elb} curves for the TH20 condition with a polynomial function (with φ_{ELB} expressed in radians). We used the Akaike Information Criterion (AIC) with correction for small sample size to match the data with a minimum of free parameters. For both elbow flexors and extensors AIC was minimal for a 3rd order polynomial:

$$T = a + b \cdot \varphi_{elb} + c \cdot \varphi_{elb}^2 + d \cdot \varphi_{elb}^3 \quad (1)$$

The values for the coefficients were averaged over the 11 participants and displayed as one set of mean a, b, c and d values and their SEM to model the T-φ_{elb} relation of the elbow flexors and one set to model the T-φ_{elb} relation of the elbow extensors.

Part of the variation found in the measured torque may be due to variation in muscle activation rather than variation in the elbow angle. To quantify the amount of variation in T–φ_{elb} curves due to angle dependent activation of the agonistic muscles we calculated Pearson's correlation coefficients between agonistic srEMG and T using all trials of the same torque direction leading to one coefficient for elbow flexors and one coefficient for elbow extensors for the whole dataset.

We wanted to establish if the T-φ_{elb} curves for the two horizontal adduction angles (TH20 and TH45) showed a shift as hypothesized. A MANCOVA was used with the elbow angle at which the curves reached their maximum (optimal elbow angle) as well as the corresponding maximal normalized torque as dependent variables and horizontal adduction angle as independent variable. Since it has been reported by Tsunoda et al. (1993) that the optimal elbow angle is different between female and male participants, we included gender as covariate.

3. Results

As an example, the torque and srEMG data recorded during the contractions of the elbow flexors in TH20 are shown in Fig. 2. The figure shows that during each contraction, there was a relatively large amount of biceps activity compared to triceps activity. The figure also suggests that the srEMG values were not constant over the different φ_{elb}, which will be discussed later.

3.1. Torque-angle curves of elbow flexors and extensors

The mean T–φ_{elb} curves for TH20 are plotted in Fig. 3. These curves had their maximum at a φ_{elb} of 95° for elbow flexors and for extensors at φ_{elb} of 55° for elbow extensors. We modeled these data with a 3rd order polynomial function (see Eq. (1)). Mean values for the coefficients a, b, c and d are given in Table 2 and the corresponding T–φ_{elb} curves are plotted in Fig. 3. For all participants

Table 1
Elbow angles as imposed on the participants (θ_{elb}) compared to the mean (SEM) elbow angles as measured (φ_{elb}) during the maximum voluntary contractions. Angles are reported in degrees with 0° being full extension in the elbow. Data of TH20 was used.

θ_{elb}	10	20	30	45	60	75	90	105	115	125
φ_{elb}, flexion	32(3)	38(2)	42(2)	57(2)	69(2)	81(2)	94(2)	105(2)	113(2)	120(1)
φ_{elb}, extension	20(2)	23(2)	31(2)	37(3)	49(3)	61(4)	78(4)	92(4)	105(5)	116(4)

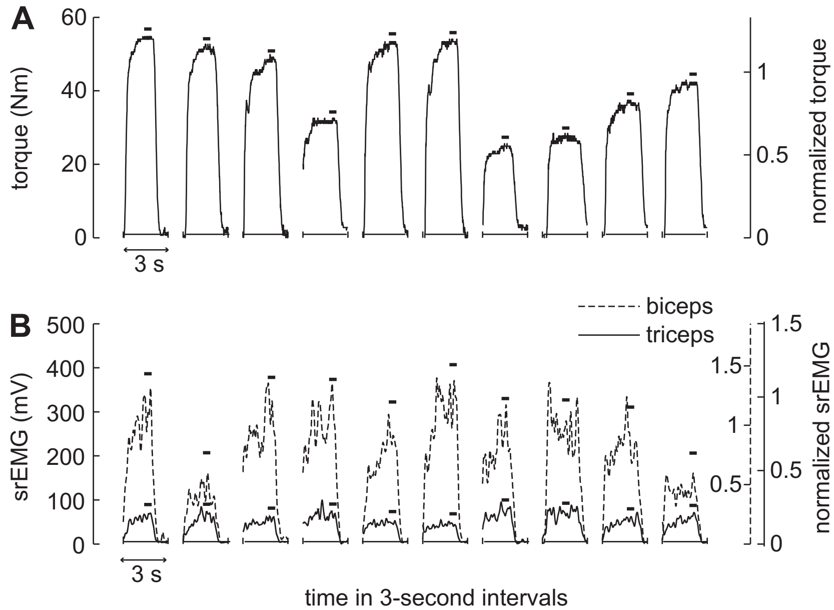

Fig. 2. Example trial of participant 9. (A) Data of one session of maximal voluntary isometric contractions of the elbow flexors. The data of the extension trials have been cut out. The time interval used to calculate the highest value per trial is marked with a block. (B) Smoothed and rectified EMG data (srEMG) of biceps and triceps. On the right side, a normalized y-axis is drawn for biceps and triceps separately.

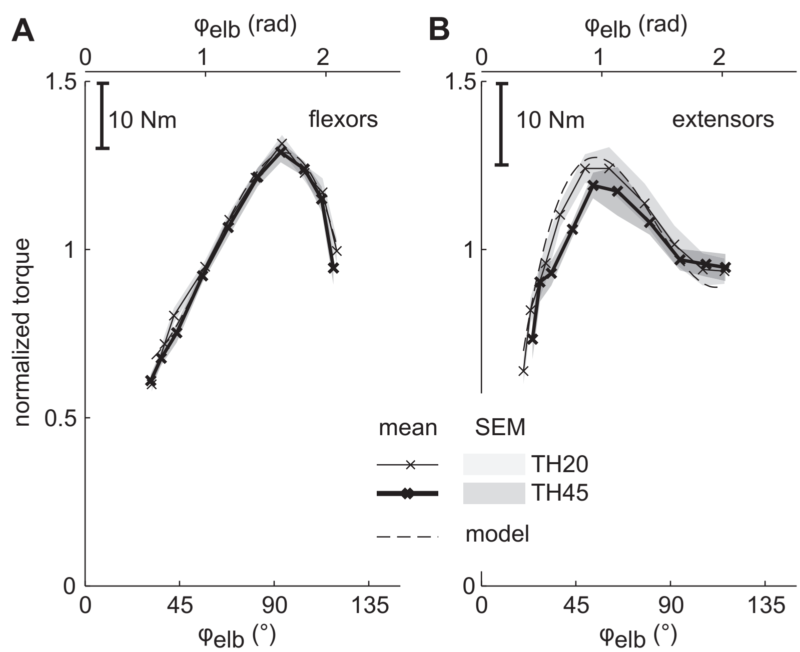

Fig. 3. Mean normalized torque (grey area: SEM) averaged over all 11 participants for each imposed elbow angle and plotted separately for a horizontal adduction angles of 20° (TH20) and 45° (TH45) in the shoulder. Curves of TH20 and TH45 show no difference in magnitude and position of the maximum value. Mean normalization factor was 50 Nm for flexion and 41 Nm for extension. The T-φ_{elb} relationship of TH20 can be described with a 3rd order polynomial function (dotted line). Coefficients of the polynomial function found for flexion torque and extension torque are presented in Table 2.

the polynomial we found explained the major part of the measured torque variance ($R^2 \geqslant 0.8$). Note that these relationships only provide a good prediction for the range of elbow angles measured in this experiment. Extrapolation to angles outside of the measurement range can lead to zero or negative torque prediction.

The maximal torque that was generated differed greatly among participants: the mean value (SEM) was 65(7) Nm for flexors and 58(8) Nm for extensors. For male participants we found a mean maximal torque of 85(6) Nm for flexors and 77(9) Nm for extensors, and for female participants we found values of 48(3) Nm for flexors and 36(2) Nm for extensors.

In order to study the effect of changing the shoulder configuration we plotted the mean T–φ_{elb} curves for TH20 and TH45 in Fig. 3. The curves are very similar. A MANCOVA on both the maximal normalized torque and optimal elbow angle showed no significant difference between TH20 and TH45 (flexors: $F_{1,19} = 0.057$, $p = 0.814$ and $F_{1,19} = 0.766$. $p = 0.392$; extensors: $F_{1,19} = 0.343$, $p = 0.565$ and $F_{1,19} = 0.093$, $p = 0.764$).

Table 2
For every participant a 3rd order polynomial function was fit on the torque–angle curves for the elbow flexors and extensors in the TH20 condition. Mean values (SEM) for the coefficients *a*, *b*, *c* and d are presented along with the mean explained variance (R^2). Note that for this fit measured elbow angles were expressed in radians.

	a	*b*	*c*	*d*	R^2
Flexors	0.896 (0.19)	−1.31 (0.58)	2.01 (0.50)	−0.648 (0.13)	0.96 (0.01)
Extensors	−0.417 (0.18)	4.27 (0.68)	−3.35 (0.67)	0.773 (0.19)	0.88 (0.02)

3.2. Angles in shoulder girdle and elbow

The realized mean φ_{th} (SEM) over all participants for the two conditions, TH20 and TH45, were 18(1)° and 44(2)° for elbow flexion, and 20(2)° and 46(2)° for elbow extension. Because the shoulder is an assembly of joints between thorax, scapula, clavicula and humerus we can divide φ_{th} into two anatomical joint: $\varphi_{ts,}$ between thorax and scapula, and φ_{sh} between scapula and humerus (see Fig. 1). In Fig. 4 these two angles, averaged over the 11 participants, are plotted against each other for each imposed elbow angle and for the two conditions (TH20 and TH45). In this figure we can see that the mean difference in φ_{th} between TH20 and TH45 ($\Delta\varphi_{th}$: 26(2)° and 26(3)° for elbow flexion and extension, respectively) was only partly reflected in a difference in φ_{sh} : 11(1)°. This means that the scapula partly moved along with the humerus and the change in φ_{th} was largely due to a change in φ_{ts}, which is not spanned by biceps and triceps. Still the φ_{sh} in TH20 showed significantly different from φ_{sh} in TH45: $t_{18} = -3.0$, $p = 0.0075$ and 9(1)°, $t_{18} = -6.2$, $p < 0.001$ for elbow flexion and extension, respectively.

3.3. EMG

Fig. 5 shows the mean srEMG-φ_{elb} curves for all conditions. Antagonistic muscle activation measured during this experiment was low and not related to elbow angle. The mean (SEM) normalized antagonistic activity was 0.26 (0.05) during flexion torque production and 0.14 (0.04) during extension torque production

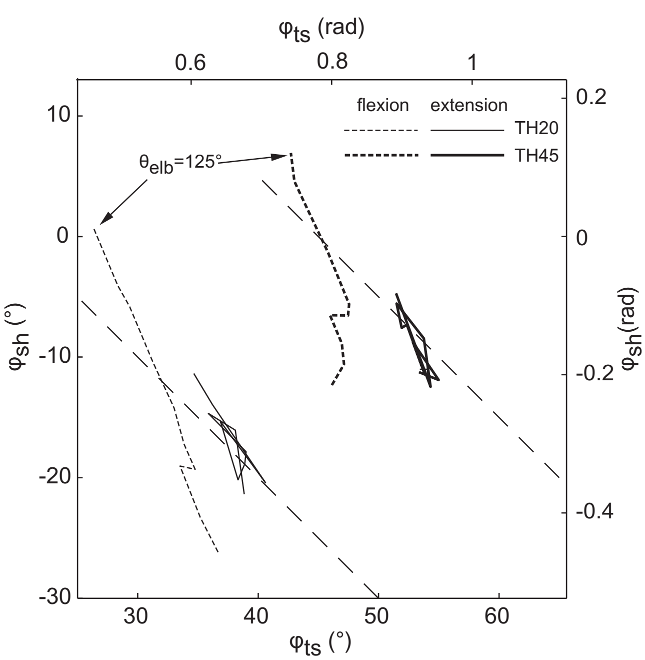

Fig. 4. Mean angle between scapula and humerus (φ_{sh}) as a function of mean angle between thorax and scapula (φ_{ts}), for a horizontal adduction angles of 20° (TH20) and 45° (TH45) in the shoulder. The long dashed lines indicate the expected relationship between φ_{sh} and φ_{ts}. Note that difference in φ_{th} between the TH20 and TH45 condition is only partly reflected in a difference in φ_{sh}.

(see Fig. 5). Agonistic muscle activation, however, was systematically related to elbow angle in a similar way as the torque of the flexors and extensors. We found a low but significant correlation between srEMG and torque: for elbow flexors (Pearson's $r = 0.21$, $p = 0.004$) and elbow extensors (Pearson's $r = 0.53$, $p < 0.001$).

4. Discussion

The purpose of this study is to gain more detailed knowledge of the isometric torque-angle relationships at the elbow, to be used in studying the control of arm movements. Firstly, we found that the curve for elbow extension reaches a maximum at an elbow angle of 50–60°. Secondly, we found no effect of changing the horizontal adduction angle on the torque-angle curve of elbow flexors and extensors. Below, we will first discuss which factors might have influenced the results. We will then compare our results on torque-angle curves with previously published data, and finally we will indicate how our results may justify a possible simplification in modeling the shoulder girdle for studies of arm movements in horizontal plane.

4.1. Factors that might have influenced the torque-angle curves of elbow flexors and extensors

An important factor potentially influencing the shape of the curve is the neural input to the muscles. In this study, we found that elbow torque correlated with measured agonistic muscle activity (See Section 3.3) with a Pearson's *r* of 0.21 for elbow flexors and 0.53 for elbow extensors. If we assume a linear srEMG-torque relationship this indicates that 5% and 28% of the variation in, respectively, measured flexion and extension torque can be explained by the variation in agonistic srEMG. This means that the greater part of the variation found in the torque-angle relationship of the elbow flexors and extensors still provides a good representation of the force-length relationships and the moment arm-angle relationships of these muscles. We can illustrate this for the elbow extensors (see Fig. 6) by separating our participants into two groups: a group of participants that showed a high correlation between agonistic srEMG and torque (group 1: $r > 0.53$), and a group of participants that showed a low correlation (group 2). In group 1, the srEMG is constant over elbow angle and the corresponding torque-angle curve, although more flattened, still shows its optimum at the same elbow angle as that of group 2. This means that for evaluation of the force-length relationship and moment arm-elbow angle relationship of the extensor muscles, the torque-angle curve of elbow extensors gives an accurate indication of the optimal elbow angle but a strong overestimation of the steepness.

Another factor potentially influencing the torque-angle relationship is activation of antagonists. We found low antagonistic activity (0.26 (0.05) during flexion torque production and 0.14 (0.04) during extension torque production) independent of elbow angle (see Section 3.3). This activation will decrease the net joint torque around the elbow as measured by the dynamometer. It is not clear how much of this antagonistic srEMG can be attributed to crosstalk. If we assume that the major part was due to activation of the antagonist and that a linear relationship between srEMG and torque was present, this means that the absolute value of the

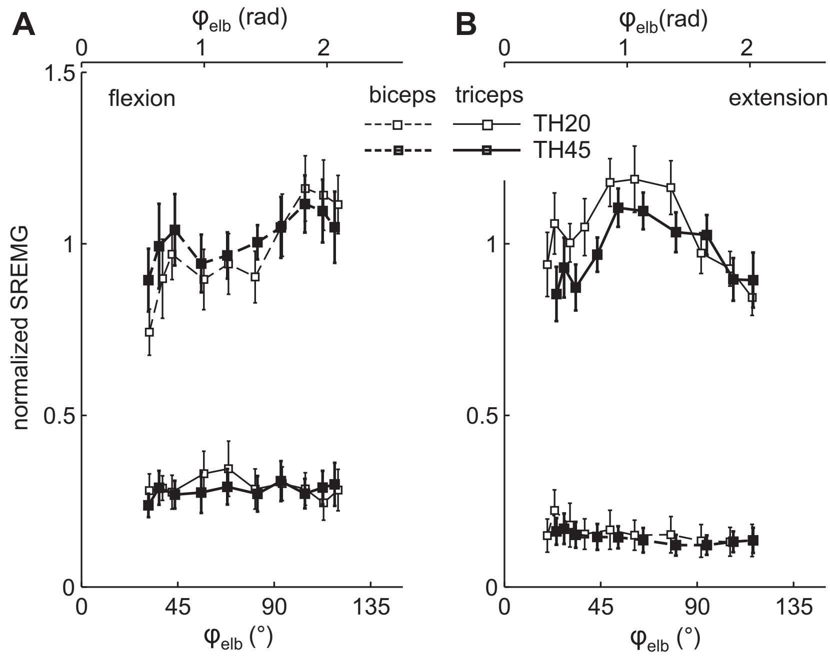

Fig. 5. Mean normalized, smoothed and rectified EMG (error bars: SEM) averaged over all 11 participants for each imposed elbow angle and plotted separately for a horizontal adduction angles of 20° (TH20) and 45° (TH45) in the shoulder. (B) A systematic relation between measured agonistic srEMG and elbow angle seems evident for elbow extension.

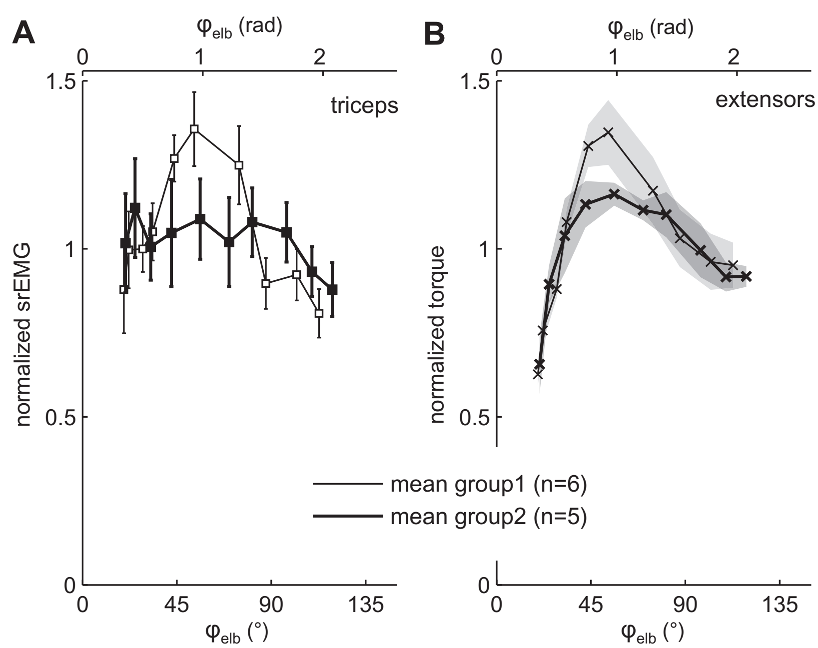

Fig. 6. For the TH20 condition, the datasets of the maximal voluntary contractions of the elbow extensors (also shown as thin lines in Figs. 5B and 3B) were divided in two groups. A group with datasets that showed a high correlation between agonistic srEMG en torque (group 1) and a group with datasets that showed a low correlation (group 2). In group 2, the mean srEMG-angle curve is almost flat and the mean torque-angle curve is flatter as that in group 1 but its maximum occurs at the same elbow angle.

maximal elbow torque produced by the extensors was underestimated by 18% and that of the flexors by 21%. In this estimate we take into account that the sum of maximal isometric torque of elbow flexors is approximately 1.25 times that of the elbow extensors (Nijhof and Kouwenhoven, 2000, and our own data, Fig. 3).

The last factor influencing the shape of the torque-angle curves of elbow flexors and extensors is the deviation in elbow angle from the imposed value (θ_{elb}) when the subject developed maximal voluntary torque. We addressed this problem by monitoring the actual elbow angle (φ_{elb}) during contractions and reported that angle instead. In Fig. 7 we plotted our torque data as a function of φ_{elb} (dotted line) and θ_{elb} (solid line). For the elbow extensors, we see that optimal elbow angle shifts to a more flexed angle when using imposed angles instead of angles measured at the elbow. Also, note that the torque-angle curves obtained using imposed angles are wider than those obtained using angles measured at the elbow and therefore overestimate torque at extreme elbow angles. Van Zuylen et al. (1988) also measured isometric torque-angle relationship for the elbow flexors using imposed elbow angles and compared them with their model predictions based on angle

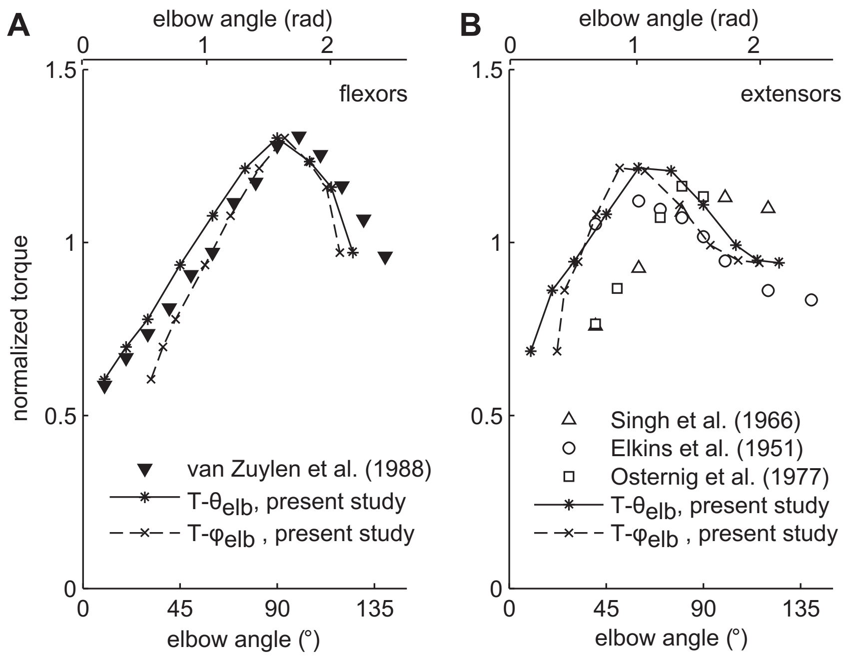

Fig. 7. Mean torque-angle curves of elbow flexors and extensors as found in the present study plotted together with results obtained in previously published studies. For the present study, data of the TH20 and TH45 condition were averaged since we found that curves showed no difference in shape between these two conditions. To facilitate the comparison of curve characteristics, all data have been normalized to their average value. The data of the present study are plotted against measured elbow angles, φ_{elb} (dashed line), and imposed elbow angles, θ_{elb} (continuous line). Data plotted with an open marker are measured in the sagittal plane.

dependent twitch torque amplitudes of the individual muscles. They found an overestimation of isometric torque-angle curve at the extreme elbow angles that is similar to our data. This confirms that using measured elbow angles during contraction gives a better indication of the width of the actual torque-angle relationships of the elbow flexors and extensors.

4.2. Isometric torque-angle curves of elbow flexors and extensors

In Fig. 7 we compared our torque-angle curve of elbow flexors and extensors with previous reported curves. Since we found similar curves for the TH20 and TH40 conditions we averaged these conditions to one curve for elbow flexors and one curve for elbow extensors. Van Zuylen et al. (1988) (Fig. 7A) measured isometric torque-angle relationships of the elbow flexors in the transverse plane with the humerus horizontally adducted by 25° and the lower arm in a neutral position (midway between pronation and supination). This position is very similar to our TH20 and we can see that the results show good agreement.

The isometric torque-angle curve of the elbow extensors showed a maximum at elbow angle of 55°. Other researchers measured torque-angle relationships in the sagittal plane and found this optimal elbow angle to occur at 60° (Elkins et al., 1951), 80° (Osternig et al., 1977) or even 100° (Singh and Karpovich, 1966), as can be seen in Fig. 7B. In the present study as well as in that of Elkins et al. (1951), elbow angle was measured at the elbow during the contraction, whereas in other studies it was measured using a goniometer attached to the dynamometer arm. If we had used imposed elbow angles (θ_{ELB}) instead of measured elbow angles (φ_{ELB}) we would have found the optimum angle at 70° (see Fig. 7B). Using this angle to compare our result with previous published work we still see that Osternig et al. (1977) and Singh and Karpovich (1966) found a more flexed angle compared to us. These studies used a set-up in which the lower arm was fixed in supination whereas in the present set-up and the set-up used in Elkins et al. (1951) the lower arm was fixed in a neutral position. Thus, the fact that other researchers find the maximal torque for elbow extensors to occur at a more flexed elbow angle then reported in this study could be due to our measuring the torque-angle relationship in the transverse plane instead of in the sagittal plane, to fixating the lower arm in a neutral position, or to both.

4.3. Isometric torque-angle curves under different horizontal adduction angles

We expected that when the humerus was adducted horizontally by 25°, short head of biceps would shorten and long head of triceps would lengthen, causing the optimum length for these muscles to shift towards elbow extension. In reality, the optimal angles found in the torque-angle curves were not affected by our intervention. First of all we should take into consideration that the contribution of the biarticular muscles to the elbow torque is small relative to that of the monoarticular elbow muscles. According to Nijhof and Kouwenhoven (2000) the sum of maximal isometric torques of the monoarticular elbow muscles is approximately 2.5 times that of the biarticular muscles. Secondly, the length changes in the biarticular muscles are probably negligible. The reason is that although we changed the horizontal adduction angle by 25°, the scapula partly followed the movement of the humerus in the transverse plane, and in fact the glenohumeral angle changed by only 10°. Based on this observation we speculate that the effect of our intervention on the length of biceps and triceps was only minimal.

In sum, the findings of this study suggest that, as long as the change of the horizontal adduction angle is small (<25°) and within the range of 20–45° relative to the thorax, the shoulder can be simplified as if the scapular bone were fixed to the humerus. This would make the biarticular biceps and triceps in the model effectively monoarticular elbow muscles. Apart from decreasing the degrees of freedom to be modeled, this has the advantage that when measuring arm movements to compare with the model simulation, the movement of the scapula does not have to be recorded. It remains to be established whether this simplification also holds when larger changes in the horizontal adduction angle are imposed.

5. Conclusions

Firstly, an isometric torque-angle relationship of the elbow extensors is now available for modeling the shoulder and elbow system in the transverse plane. Secondly, for the most part the observed variation in normalized extensor torque with elbow angle can be attributed to force-length relationship and the moment arm-angle relationship of the elbow extensors. Thirdly, we recommend that for future measurement of torque-angle relationships actual joint angles during contraction be measured to avoid overestimation of the joint torque at extreme elbow angles. Fourthly, we conclude that isometric torque-angle relationships at the elbow in the transverse plane are not influenced by moderate changes in horizontal adduction angle of the humerus.

Acknowledgement

We thank the Rehabilitation Center Amsterdam for allowing us to conduct our measurements in the Duyvensz-Nagel Research Laboratory.

Appendix A

In this appendix, we will clarify how the measurement error due to misalignment between the rotation axis of the dynamometer and the rotation axis of the elbow was estimated.

The net joint torque produced around the elbow (T_E) can be expressed as:

$$T_E = |EP| \cdot F_P \tag{A1}$$

With $|EP|$ the distance between the rotation axis of the elbow and the point of force transduction (P) from the hand to the handle of the dynamometer, and F_{TE} the force applied at P perpendicular to the lower arm. When assuming that all force generated by the participant was applied perpendicular to the lower arm (as was instructed to the participant), torque measured by dynamometer (T_R) can be expressed as:

$$T_R = |RP| \cdot F_{P,y} = |RP| \cdot F_P \cdot \cos(\alpha) \tag{A2}$$

With $|RP|$ the distance between the rotational axis of the dynamometer (R) and P and α the angle of misalignment (see Fig. A1). We define the relative measurement error relative to T_E as:

$$\text{Error} = \frac{T_R - T_E}{T_E} \tag{A3}$$

This error can be rewritten as:

$$\text{Error} = \frac{|RP| \cdot F_P \cdot \cos(\alpha) - |EP| \cdot F_P}{|EP| \cdot F_P} \tag{A4}$$

As the set-up was adjusted to the participant so that E was inline with R this means that $|RP| = |EP|$. When assuming that the misalignment does not affect $|EP|$ we can rewrite Eq. (A4) to a simple relation between the measurement error and the angle of misalignment:

$$\text{Error} = \cos(\alpha) - 1 \tag{A5}$$

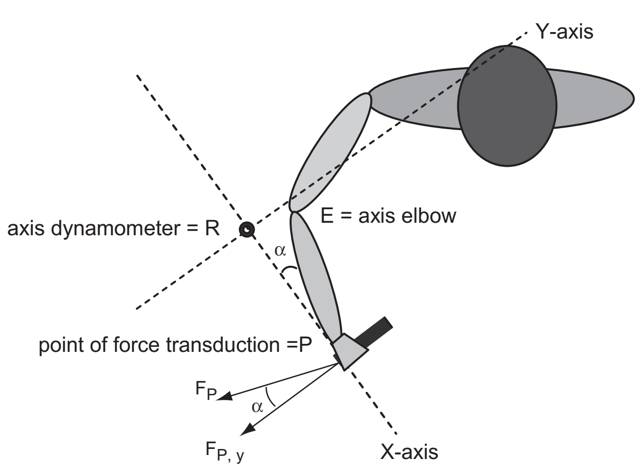

Fig. A1. The misalignment between the rotational axis of the elbow and the axis of the dynamometer can be expressed as the angle between the participant's lower arm and the dynamometer arm (α). The relative error in the measured torque as a function of α was estimated. It was assumed that participants applied force at P (F_P) perpendicular to their lower arm.

References

Brondino L, Suter E, Lee HD, Herzog W. Elbow flexor inhibition as a function of muscle length. Journal of Applied Biomechanics 2002;18(1):46–56.
Dowling JJ, Konert E, Ljucovic P, Andrews DM. Are humans able to voluntarily elicit maximum muscle force. Neuroscience Letters 1994;179(1–2):25–8.
Elkins EC, Leden UM, Wakim KG. Objective recording of the strength of normal muscles. Archives of Physical Medicine and Rehabilitation 1951;32(10):639–47.
Gomi H, Kawato M. Human arm stiffness and equilibrium-point trajectory during multi-joint movement. Biological Cybernetics 1997;76(3):163–71.
Gribble PL, Mullin LI, Cothros N, Mattar A. Role of cocontraction in arm movement accuracy. Journal of Neurophysiology 2003;89(5):2396–405.
Karduna AR, McClure PW, Michener LA, Sennett B. Dynamic measurements of three-dimensional scapular kinematics: a validation study. Journal of Biomechanical Engineering 2001;123(2):184–90.
Kistemaker DA, Van Soest AKJ, Bobbert MF. Is equilibrium point control feasible for fast goal-directed single-joint movements? Journal of Neurophysiology 2006;95(5):2898–912.
Meskers CGM, van de Sande MAJ, de Groot JH. Comparison between tripod and skin-fixed recording of scapular motion. Journal of Biomechanics 2007;40(4):941–6.
Nijhof EJ, Gabriel DA. Maximum isometric arm forces in the horizontal plane. Journal of Biomechanics 2006;39(4):708–16.
Nijhof EJ, Kouwenhoven E. Simulation of multijoint arm movements. In: Biomechanics and neural control of posture and movement. Springer; 2000. p. 363–72.
Osternig LR, Bates BT, James SL. Isokinetic and isometric torque force relationships. Archives of Physical Medicine and Rehabilitation 1977;58(6):254–7.
Singh M, Karpovich PV. Isotonic and isometric forces of forearm flexors and extensors. Journal of Applied Physiology 1966;21(4):1435–7.
Smeets JBJ, Erkelens CJ, Dernier van der Gon JJD. Adjustments of fast goal-directed movements in response to an unexpected inertial load. Experimental Brain Research 1990;81(2):303–12.
Tsunoda N, Ohagan F, Sale DG, Macdougall JD. Elbow flexion strength curves in untrained men and women and male bodybuilders. European Journal of Applied Physiology and Occupational Physiology 1993;66(3):235–9.
Van Zuylen EJ, Van Velzen A, Dernier van der Gon JJ. A biomechanical model for flexion torques of human arm muscles as a function of elbow angle. Journal of Biomechanics 1988;21(3):183–90.

Ilona J. Pinter graduated as Exercising Therapist in 1994 at the School for Exercise Therapists Mensendieck in Amsterdam, The Netherlands. She worked as a registered medical therapist at her own practice, in Amsterdam until 2002. She obtained her M.Sc. degree in Human Movement Sciences at the VU University Amsterdam, the Netherlands, in 2006. Currently she works at this faculty as a Ph.D. student and investigates the control of fast point-to-point arm movements in both experimental set-up and simulation study.

Maarten F. Bobbert obtained in 1983 a M.Sc. degree in Functional Anatomy as well as a M.Sc. degree in Exercise Physiology from the Faculty of Human Movement Sciences of the VU University Amsterdam, The Netherlands. At this faculty, he also conducted his Ph.D. work on muscle functioning and coordination in vertical jumping, which was finalized in 1988. After a two-year post-doc appointment at the Human Performance Laboratory of the University of Calgary, Canada, Maarten returned to Amsterdam to be appointed as staff member. His current teaching includes courses on neurosciences and neuro-musculo-skeletal modeling. Among his research interests are the design of the musculo-skeletal system and the organization of control in the central nervous system. He is editorial consultant of the Journal of Biomechanics, and editorial board member of Human Movement Science and Clinical Biomechanics.

Knoek van Soest obtained his Ph.D. in 1992 from the Faculty of Human Movement Sciences, VU University Amsterdam, where he has worked ever since. His research focuses on the mechanics, energetics and control of multi-joint human movements, using a combination of experimental methods and computer simulation and optimization. He is currently the chairperson of the ISB Technical Group on Computer Simulation that organizes a highly interesting biannual symposium in conjunction with the ISB conference. He is an editorial board member of Medicine and Science in Sports and Exercise.

Jeroen B.J. Smeets obtained in 1987 a M.Sc. degree in Experimental Physics at Utrecht University. At the same university, he obtained in 1991 a Ph.D. based on research on reflex coordination and the control of arm movements. Subsequently he became a postdoc and later staff member at the department of Neuroscience of the Erasmus MC, Rotterdam. In 2006, he became full professor at the Faculty of Human Movement Sciences of the VU University Amsterdam. His current teaching includes an undergraduate course on motor coordination and master courses on behavioral concepts in human movement sciences and on perception in action. His research spans a broad field sensorimotor control, ranging from visual and haptic perception to motor control. He is editor of Acta Psychologica, and editorial board member of Journal of Motor Behavior.